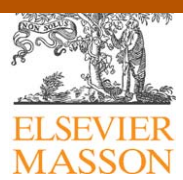
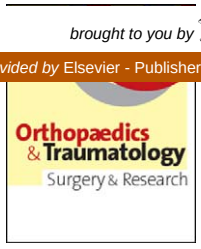
Orthopaedics & Traumatology: Surgery & Research (2010) 96, 646—651



Available online at ScienceDirect
www.sciencedirect.com

Elsevier Masson France
EM|consulte
www.em-consulte.com



ORIGINAL ARTICLE

Fiber Bragg grating sensor for monitoring bone decalcification

V. Mishra [a,*], N. Singh [a], D.V. Rai [b], U. Tiwari [a], G.C. Poddar [a], S.C. Jain [a], S.K. Mondal [a], P. Kapur [a]

[a] *Central Scientific Instruments Organisation, Council of Scientific and Industrial Research, Sector-30, Chandigarh 160030, India*
[b] *Center for Biomedical Engineering, Institute of Engineering and Technology, Shobhit University, Meerut 250110, India*

Accepted: 19 April 2010

KEYWORDS
Fiber Bragg Grating;
Decalcification;
Bending strain

Summary
Introduction: Estimation of decalcification is a vital tool to discern bone health. Different techniques are used for its quantitative measurement, e.g. DEXA, QCT & QUS. All these techniques, although noninvasive, suffer from limitations such as radiation exposure and inaccurate values. Recently, fiber optic techniques are fast emerging for medical applications owing to their various attractive features like immunity to EMI/RFI, geometric versatility, chemical inertness, etc.
Material and methods: The effect of decalcification on strain response of a goat tibia was investigated *in vitro* using fiber Bragg grating (FBG) sensing technique. The bone was strained by using three-point bending technique and corresponding Bragg wavelength shifts were recorded. Two similar bone samples from the same animal were taken and one was partially decalcified. Strain response of decalcified and untreated bone was taken concurrently to monitor the effects of calcium loss and that of degradation with time.
Results and conclusion: The strain generated for same stress increased with greater degree of decalcification and a steep increase occurred after 2 g calcium loss, indicating the onset of damage. The strain response, therefore gives a direct indication of the degree of calcium present in the bone.
Level of evidence: Level III.

Introduction

Optical fibers offer many advantages for biomedical applications [1] due to their well-known intrinsic properties such as small size (thickness less than that of standard surgical suture), biocompatibility, non-toxicity, immunity to electro-

* Corresponding author. Tel.: +0091 172 2659951; fax: +0091 172 2659951/2657082/267.
E-mail address: vandanamishra66@yahoo.com (V. Mishra).

1877-0568/$ - see front matter © 2010 Elsevier Masson SAS. All rights reserved.
doi:10.1016/j.otsr.2010.04.010

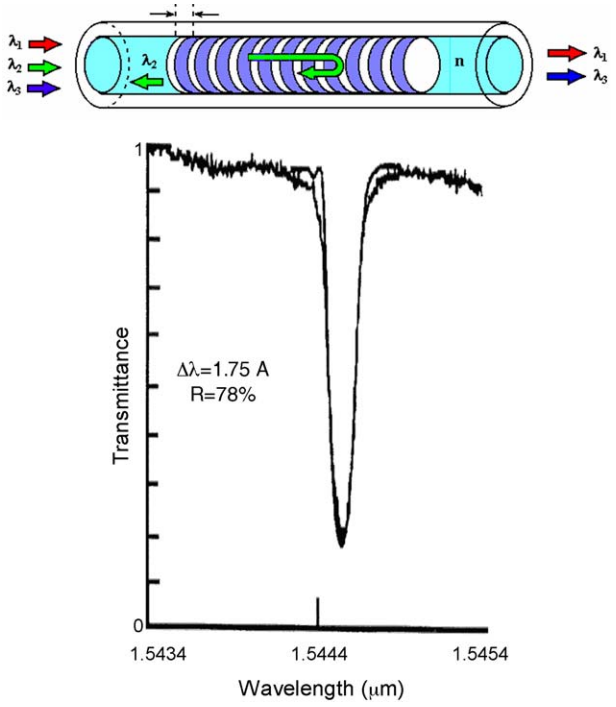


Figure 1 Schematic of the fiber Bragg grating and its transmission spectrum.

magnetic and radio frequency radiations as well as chemical inertness. As they are intrinsically safe for the patient, they can be used for *in vivo* measurements and can be left in their position for repeated or continuous monitoring. These qualities allow them to be safely used without any interference in a clinical setup. Amongst various fiber-sensing technologies, fiber Bragg grating (FBG) are intrinsic fiber devices, which function by controlling the properties of light propagating in a photosensitive fiber core. This technology is fast emerging because it offers all the features of optical fibers with some added advantages as well, for example, self-referencing as the information is wavelength encoded and ease of multiplexing facilitating distributed sensing [2]. In a fiber grating there are no ''lines'' or ''grooves'' etched on the surface. Instead, they are uniformly spaced points in a fiber core where the refractive index has been raised from that of the rest of the core by illuminating it with UV light as shown schematically in Fig. 1. If light from a broadband source is transmitted through such a fiber, one particular wavelength is scattered and will be missing from the transmission spectrum. The potential of FBGs as strain and temperature sensors array in various concrete and composite structures for health monitoring had been established much earlier [3—6], but in the biomedical arena this technology is still under research and development stage in spite of the fact that their multiparameter and minimally invasive sensing capabilities make them highly suitable for medical applications.

For *in vivo* applications, FBGs have an edge over conventional gauges because they have smaller risk for infection and can also be used even on curved surface or in locations where the use of conventional gauge is technically and medically not feasible. For example, conventional electrical strain gauges (ESGs), considered gold standard for strain measurement, consist of metallic parts and are difficult to adhere on the bone surface. Besides, working of an ESG depends upon measuring electric resistance that varies with applied strain and so it is not suitable in strong electric and magnetic field environment associated with medical appliances. Moreover, they cannot be made completely biocompatible. Various uses of FBGs as temperature and distributed pressure sensor gauges have been explored in the field of medicine [7—9]. An FBG strain sensor has been proposed by a Brazilian group for monitoring ventilatory movements for patients in ICU [10]. Use of FBG sensors in dental biomechanics had also been investigated earlier [11]. An embedded array of FBGs can be used for pressure mapping at different orthopedic joints. The researchers at Nanyang Technological University, Singapore have reported an FBG based sensor in instrumented tibial spacer to correct misalignment during total knee replacement surgery. The sensor constitutes an FBG array embedded in fiber-reinforced composite [12]. Recently some studies on the possible use of FBG as strain gauge in bones have been undertaken. Although *in vivo* strain measurements in human is not very common, the researchers at Hadassah University Hospital, Israel have reported the use of instrumented bone staples made of electrical strain gages in some volunteers for the same [13—15]. Talaia et al. [16] have first reported the use of FBG sensor array to study strains in fracture fixation of synthetic femur. Fresvig & coworkers have validated the use of FBG sensors in place of ESGs to measure deformation in human cadaver femur bone specimen under *in vitro* loading condition [17]. The present study takes these concepts further by proposing the use of FBG sensors in place of instrumented bone staples to evaluate strain for assessment of mineral loss of a goat bone sample through *in vitro* measurements.

Calcium is the most important mineral in body and decalcification can change the mechanical integrity of the bones. The extent of decalcification is conventionally assessed by bone mass density (BMD) measurement and comparing it with that of a normal healthy sample. Frequently used techniques for densitometry include dual energy X-ray absorptiometry (DEXA), quantitative computer tomography (QCT) and quantitative ultrasound assessment (QUS). All these techniques, although noninvasive, suffer from various limitations such as radiation exposure, inaccurate values, e.g. they may be falsely elevated in the presence of extensive degenerative change, aortic calcification, or vertebral compression fractures, presence of spinal rods or hip replacements [18]. Moreover, corrective measures are needed for difference in age, gender, ethnicity, height and weight from patient to patient. In the present experimental investigation, FBG based sensing technique was explored for this assessment by monitoring changes in the mechanical loading properties of bone through *in vitro* strain measurements of chemically decalcified sample with respect to that of untreated one. In this technique, there is no radiation exposure involved and the possibility of error is minimized as it gives self-referenced values. Additionally, since FBGs can detect microstrain values, it is possible to detect decalcification just at the onset of the damage and thereby helps in advance alarming of bone degeneration. The objective of the present work was to establish an FBG based method for monitoring various degrees of decalcification which has

not been reported till date to the best of our knowledge. This was accomplished by using FBG technology to detect decalcification by monitoring strain response of chemically treated and untreated bone under same conditions.

Working principle

When a light wave enters a medium with varying refractive indices, it undergoes minute reflections from every interface. If all the individual reflections are in phase, constructive interference will take place between reflected waves leading to strong reflection at a particular wavelength given by Bragg equation,

$$\lambda_B = 2n\Lambda$$

where λ_B is reflected Bragg wavelength, n is the effective refractive index of the core and Λ is the pitch of the grating. Therefore, when light from a broadband source is launched in an FBG, the spectral component defined by the above equation is missing from the transmitted spectrum (Fig. 1). Bragg wavelength is shifted if the effective refractive index or the grating periodicity is changed due to some perturbation; in fact both these parameters are directly influenced by strain and ambient temperature with the associated wavelength shift given as,

$$\Delta\lambda_B = 2\left[\Lambda\frac{\partial n}{\partial l} + n\frac{\partial \Lambda}{\partial l}\right]\Delta l + 2\left[\Lambda\frac{\partial n}{\partial T} + n\frac{\partial \Lambda}{\partial T}\right]\Lambda T$$

where Δl is the change in grating length due to strain and ΔT is the change in ambient temperature. The first term on the RHS gives strain dependence while the second term gives temperature dependence of the Bragg wavelength. A standard FBG with Bragg wavelength $\sim$1550 nm has a strain sensitivity of 1.2 pm/$\mu\varepsilon$ at constant temperature and temperature sensitivity of 12 pm/°C at zero strain [2,5].

To test the mechanical strength of a bone sample it is very important to simulate as closely as possible the straining actions taking place in the living beings in real-life situations. There are four types of straining actions possible namely, axial compression, axial tension, bending and twisting. Of these four types, bending test was undertaken in this investigation because it is the action to which bone is normally subjected either by muscular pull, by the weight of the body or by accidental violence [19]. Also, bending is a convenient method of applying a large mechanical load which gives an easily observable deformation or strain.

Materials and methods

For this investigation, two FBGs were designed and fabricated by exposing the core of a photosensitive fiber (Stocker Yale Inc.) to intense UV light from a KrF excimer laser at 248 nm through a phase mask of period 1060 nm [20]. The Bragg wavelengths of these FBGs were 1551.5 and 1550.8 and lengths were 1 cm each. Two tibia bone samples were taken from same animal. They were cleaned and surface prepared for attaching FBG sensors. The sensor was directly bonded onto the surface at the midpoint of the bone shaft,

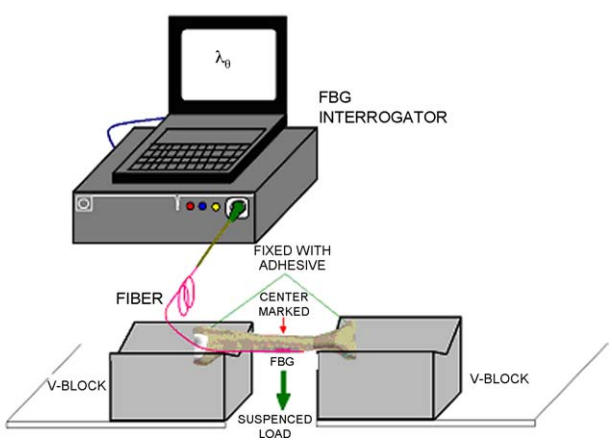


Figure 2 Schematic of the experimental setup.

which is the most vulnerable point, using standard cynoacrylate adhesive, known to work well with body tissues [21] and cured properly. One sample was decalcified in steps while the other was kept in saline solution for a comparative study. Both the samples were kept in saline solution in the refrigerator when not in use. For three-point bending test, a mechanical setup was established in a configuration similar to that reported by Bell et al. [19]. The bone was held and fixed in a pair of identical V-grooves as represented schematically in Fig. 2, so that the position of the sensor and the point of load application remain unchanged throughout the experiment. Fig. 3 is the photograph of the actual setup used in the investigation. The samples was stressed by applying increasing load from 100 g to 4 kg and the corresponding Bragg wavelength shift was monitored using an interrogator (Si 425, Micron optics). Strain values were calculated from the Bragg wavelength shift using the standard calibration factor.

One of the bone samples was then removed from the groove and soaked in 5% calcium chelating solution (CCS) [22] for 2 hours (treatment 1); it was then removed and washed thoroughly with double distilled water. The used chelating solution was analyzed to detect the presence of calcium and other minerals. Strain investigations were

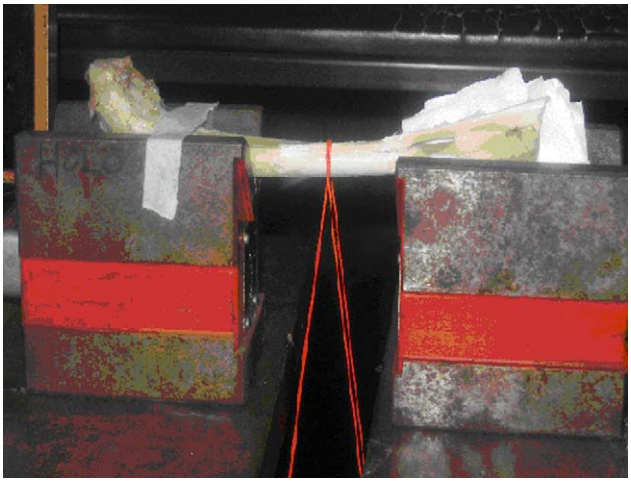


Figure 3 Photograph of the experimental setup.

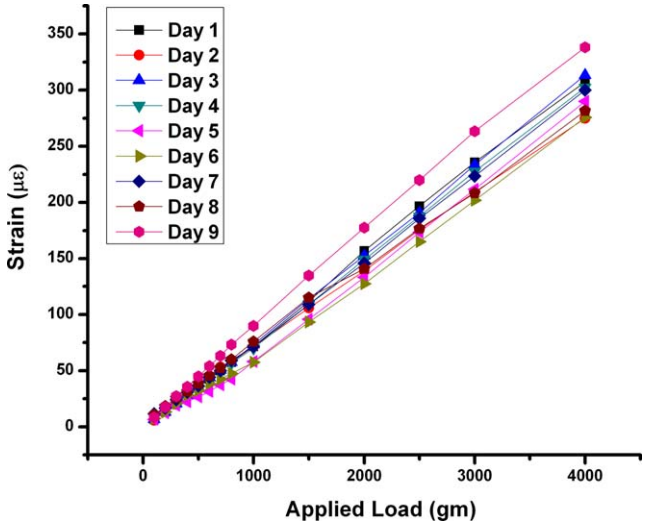


Figure 4 Strain response of untreated bone.

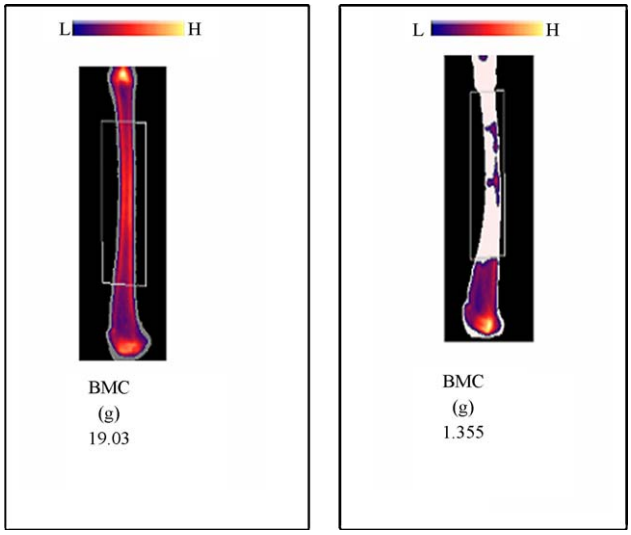


Figure 5 DEXA results of untreated and treated bone.

repeated for the treated sample as well as untreated sample, carefully reproducing the same arrangement of sensor as well as of load. In further treatments, the same procedure was repeated with increased soaking time, i.e. the sample was soaked for 3, 4, 7, 10 14, 16 and 18 hrs in that order, after which the sensor got damaged. The strain response of untreated bone was repeated each time side-by-side for comparison and to monitor the effect of time. The whole experimental procedure was spanned through a time period of 9 days, after which both the samples were analyzed using conventional DEXA technique to validate decalcification process as well as to evaluate total mineral loss.

Results

The strain response of the untreated bone remained largely unchanged throughout the time period of experiment, i.e. 9 days (Fig. 4). An average value of 76.2 microstrain per kg with a variation of ± 4.75 was observed. This slight variation can be attributed to the relative dryness of the sample at the time of strain measurement. For the treated bone, the strain response increased from the control value of 57.5 microstrains per kg with increasing decalcification. There was no measurable change in the dimensions (length, diameter) of both the samples, while the weight of the decalcified sample was reduced more with each treatment as compared to the normal bone as shown in Table 1.

After completion of the experiment, weight of untreated bone was reduced by 7.5 g (6.7%) while that of treated bone was reduced by 29 g. If 6.7% weight loss due to time and other factors such as moisture loss is taken into account, the weight loss due to demineralization is 21.6 g. Analysis of the chelating solutions after each decalcification treatment gave cumulative mineral loss of ~21.7 g which matches well with the bone-weight loss. According to DEXA reports total mineral loss of the sample is close to 17.7 g, assuming both the samples have similar mineral composition.

Discussion

The difference in DEXA value and the value obtained experimentally may be because of limited bone-area exposed to DEXA, as evident from Fig. 5, which shows the DEXA images and value of bone mineral content (BMC) values for both the samples after the experiment. Calcium loss in each case was

Table 1 Comparison of mechanical properties of untreated and treated bone.

	Untreated bone		Treated bone			
	Strain grad. (με/kg)	Weight (g)	Strain grad. (με/kg)	Weight (g)	Cumulative Ca loss (g)	Cumulative mineral loss (g)
Day 1/No Treatment	78.6	112.93	57.50	109.38		
Day 2/Treatment 1	75.0	109.06	71.66	104.85	0.3906	2.7781
Day 3/Treatment 2	80.0	108.13	83.33	102.12	0.7080	4.9696
Day 4/Treatment 3	77.5	106.81	86.40	103.5	1.0678	7.9918
Day 5/Treatment 4	77.2	105.99	93.07	97.77	1.4062	10.5984
Day 6/Treatment 5	72.8	105.95	91.10	88.30	1.8671	14.638
Day 7/Treatment 6	66.1	105.56	220.60	89.20	2.2682	17.925
Day 8/Treatment 7	76.1	105.37	921.9	83.30	2.6094	20.6057
Day 9/Treatment 8	82. 8	105.36	1309.2	80.38	2.7840	21.7025

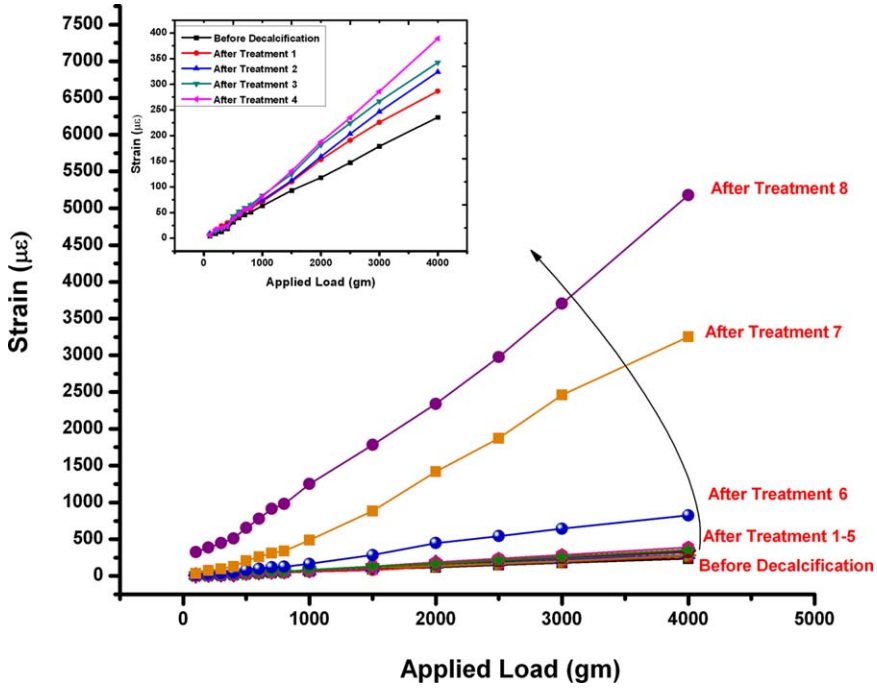


Figure 6 Strain response of chemically treated bone.

assessed by chromatographic analysis of the corresponding CCS'.

As for treated bone, after chemical treatment, the strain value for the same load increased. This increase became higher with more decalcification (Fig. 6). Strain per unit load or strain gradient is reflection of elastic property of materials as elasticity is inversely proportional to strain, thus more mineral loss results in lowering the stiffness of the bone. Strain was found to increase for the same load as the degree of bone decalcification increased by recurring treatments. Strain gradient was calculated using Figs. 5 and 6. As shown in the Fig. 7 up to 2 g of cumulative calcium reduction this increase was almost linear after that a steep and non-linear change was observed. It is obvious that this amount of calcium loss is threshold for bone damage; a slight load can result in very large strain. For example, calcium loss of even 0.3906 g (treatment 1) resulted in 1.3 times/24% more strain for same load and a calcium loss of 1 g resulted in 50% increase in strain. As the calcium loss was more than 2 g, the strain increase was close to 300% and after treatment 8 when calcium loss was 2.78 g the strain increase reached more than 2000% i.e. 22 times more strain as compared to strain before decalcification.

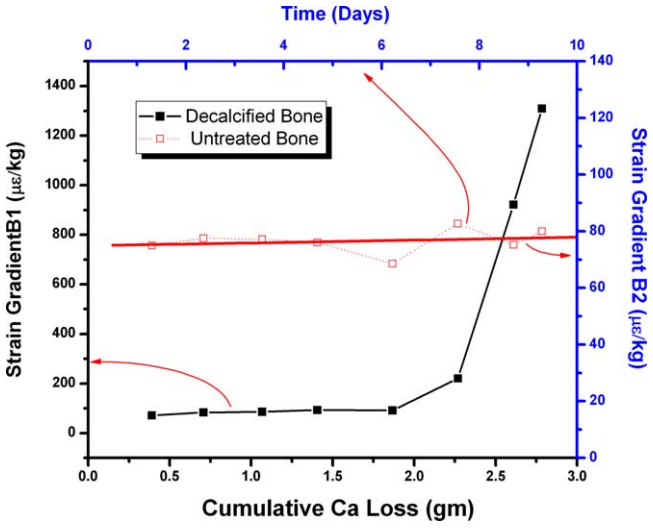


Figure 7 Comparison of strain response of treated and untreated bone.

This indicates that the modulus of elasticity or stiffness of the sample was decreased due to decalcification giving rise to reduced bone strength. It was also observed that, when the load was removed in case of untreated bone, it attained its original position immediately while for the treated bone sample; the restoration time was noticeably longer, increasing with the level of decalcification. After treatment 6, when calcium loss reached 2 g, the restoration time was 28 min while after treatment 7, it was close to 6 hrs and finally after next treatment, the bone got deformed permanently during strain measurement and the sensor got damaged. Since elasticity decreased considerably because of decalcification, the treated sample started showing more and more plastic behavior for a lower stress (load) and after treatment 8, it was completely plastic.

It is possible to measure strain less than five microstrains accurately using this sensing technique, which can be indicative of the onset of decalcification. Since the experiments were performed in laboratory, ambient temperature was constant; there was no need for temperature compensation. Nevertheless, for real implementation of these sensors, this cross-sensitivity can be avoided using suitable compensating technique [23,24]. It is to be mentioned here that as bone is not an isotropic material, for the assessment of its real mechanical competence with respect to applied

load, measurement of elastic properties at multiple directions and at various locations is necessary. However, in the present investigation, the focus was to monitor the effect of decalcification on the elastic properties of bone and the strain response was monitored at a particular location under the same experimental conditions. The aim of this work was not to offer a comprehensive description of the changes in mechanical properties of the bone but to evaluate if this FBG sensor technology can be used effectively to determine changes in same due to various degrees of decalcification and to predict if these sensors have a role in such estimation carried out by biomedicine experts. Experiments were performed for *in vitro* measurements but it can be easily adapted for *in vivo* measurements as well with some advancements and innovative modifications as per requirement especially for athletes, military persons and astronauts. The small size of fiber can be utilized to make strain staples much smaller then the existing ones [13—15] so that it can be implanted using *minimally invasive* surgical method, or, as this technology is still developing it may advance into a *noninvasive* method in the future.

Conflict of interest statement

None.

Acknowledgements

The authors acknowledge the financial support provided by the Council of Scientific and Industrial Research (CSIR) for the present work that was carried out under the Supra Institutional Project of CSIO/CSIR and Department of Radio-Diagnosis & Imaging, Postgraduate Institute of Medical Education & Research (PGIMER), Chandigarh for DEXA analysis of the bone samples.

References

[1] Mishra V, Singh N, Kapur P. Fiber optic sensors technology in medicine. In: Rai DV, Raj Bahadur, editors. Trends in medical physics and biomedical instrumentation. India: New Era International Imprint; 2009. p. 225—42.
[2] Kashyap R. Fiber Bragg gratings. London, UK: Academic Press; 1999.
[3] Othonos A, Kalli K. Fiber Bragg gratings, fundamentals and applications in telecommunications and sensing. Norwood, MA: Artech House INC; 1999.
[4] Rao YJ. Recent progress in applications of in-fiber Bragg grating sensors. Opt Lasers Eng 1999;31:297—324.
[5] Schulz WL, Udd E, Seim JM, McGill GE. Advanced fiber grating strain sensor systems for bridges, structures and highways 1998; www.BlueRR.com.
[6] Singh N, Jain SC, Mishra V, Poddar GC, Jindal VK, Bajpai RP, et al. Multiplexing of fiber Bragg grating sensors for strain and temperature measurements. Exp Tech 2007;31(3):54—6.
[7] Rao YJ, Webb DJ, Jackson DA, Zhang L, Bennion I. In-fiber Bragg grating temperature sensor for medical applications. J Lightwave Technol 1997;15(5):779—85.
[8] Dennison CR, Wild PM, Byrnes PWG, Saari A, Itshayek E, Wilson DC, et al. *Ex vivo* measurement of lumbar intervertebral disc pressure using fiber-Bragg gratings. J Biomech 2008;41(1):221—5.
[9] Milczeswki MS, Silva JCC, Simões JA, Paterno AS, Kalinowski H. Measurement of orthodontic forces and tooth displacement using fiber Bragg grating sensors. In: 5th International Conference on Mechanics Materials and Design. 2006.
[10] Wehrle G, Nohama P, Kalinowski H, Torres PI, Valente LCG. A fiber optic Bragg grating strain sensor for monitoring ventilatory movements. Meas Sci Technol 2001;12: 805—9.
[11] Carvalho L, Silva JCC, Nogueira R, Pinto JL, Kalinowski H, Simões JA. Application of Bragg grating sensors in dental biomechanics. J Strain Anal 2006;41(6):411—5.
[12] Mohanty L, Tjin SC, Lie D, Panganiban SEC, Chow P. Fiber grating sensor for pressure mapping during total knee arthroplasty. Sens Actuators A Phys 2007;135(2):323—8.
[13] Milgrom C, Finestone A, Simkin A, Ekenman I, Mendelson S, Millgram M, et al. *In vivo* strain measurements to evaluate the strengthening potential of exercises on the tibial bone. J Bone Joint Surg Br 2000;82-B:591—4.
[14] Milgrom C, Finestone A, Levi Y, Simkin A, Ekenman I, Mendelson S, et al. Do high impact exercises produce higher tibial strains than running? Br J Sports Med 2000;34:195—9.
[15] Milgrom C, Finestone A, Segev S, Olin C, Arndt T, Ekenman I. Are overground or treadmill runners more likely to sustain tibial stress fracture? Br J Sports Med 2003;37:160—3.
[16] Talaia PM, Ramos A, Abe I, Schiller MW, Lopes P, Nogueira RN, et al. Plated and intact femur strains in fracture fixation using fiber Bragg gratings and strain gauges. Exp Mech 2007;47:355—63.
[17] Fresvig T, Ludvigsen P, Steen H, Reikeras O. Fiber optic Bragg grating sensors: an alternative method to strain gauges for measuring deformation in bone. Med Eng Phys 2008;30: 104—8.
[18] Syed Z, Khan A. Bone densitometry: applications and limitations. J Obstet Gynaecol Can 2002;24(6):476—84.
[19] Bell GH, Cuthbertson DP, Orr J. Strength and size of bone in relation to calcium intake. J Physiol 1941;100: 299—317.
[20] Singh N, Jain SC, Aggarwal AK, Bajpai RP. Fiber Bragg grating writing using phase mask technology. J Sci Ind Res 2005;64:108—15.
[21] Seifman BD, Rubin MA, Williams AL, Wolf JS. Use of absorbable cyanoacrylate glue to repair an open cystotomy. J Urol 2002;167(4):1872—5.
[22] Saha S, Williams P, Rai DV, Albright JA. Electrical properties of demineralised bone. In: Digest of Papers, 8th Southern Biomedical Engineering Conference. 1989.
[23] Pant J, Mitra A, Tiwari U, Mondal S, Singh N, Jain SC, et al. Temperature compensated transverse load sensor based on dual FBG sensor. In: (IEEE Explorer) Proceedings of International Conference on Microwave — 08. 2008. p. 395—7.
[24] Zhou Z, Ou J. Techniques of temperature compensation for FBG strain sensors used in long-term structural monitoring. Proc SPIE 2005;5851:167—72.